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SENSORY INFORMATION FROM AFFERENT NEURONS

Contract No.: NIH-NINDS-NO1-NS-6-2339

PROGRESS REPORT #4

for the period

1 May 1997 to 30 September 1997

Principal Investigator: J.A. Hoffer, Ph.D.

Research Associates: Y. Chen, Ph.D.
K. Strange, M.A.Sc.
F. Maloufi, D.V.M.
T. Blasak, R.L.A.T.

Graduate Students: P. Christensen, M.A.Sc.
D. Crouch, M.Sc.
E. Heygood, B.Eng.

Neurokinesiology Laboratory
School of Kinesiology
Simon Fraser University,
Burnaby, B.C., V5A 1S6, Canada

Subcontractors: B.J. Andrews, Ph.D.
A. Kostov, Ph.D.

Faculty of Rehabilitation Medicine,
University of Alberta
Edmonton, Alberta, T6G 2G4, Canada

R. B. Stein, D.Phil.
K. Yoshida, Ph.D.

Neuroscience Division, University of Alberta
Edmonton, Alberta, T6G 2G4, Canada

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I. Objectives of Overall Project

Our aim is to develop and perfect, in an animal model, methods for chronic recording and processing of afferent activity produced by sensory receptors that could yield information about human fingertip contact, grasped object slip, finger position, and grasp force applicable for restoration of motor functions in the paralyzed human hand. The specified contract objectives are:

1. Select recording methods that:
   a. Have the potential of providing safe, reliable recordings in humans for periods of years.
   b. When used in human applications, could provide relatively isolated information from the sensory endings in the thumb pad and in the finger pads of the second and third fingers.
   c. Could, in human applications, provide information from the proprioceptive receptors in the muscles of the hand and wrist.

2. Select an animal model suitable for chronic recording of afferent nerve activity, and give consideration to modeling electrode placement sites for a potential human neural prosthesis application.

3. Fabricate or obtain chronic electrodes and associated cables and percutaneous connectors for chronic recording of sensory afferent activity.
   a. Design electrodes and cables using biocompatible materials that would be suitable for potential future human implants.
   b. Design electrodes and cables with the goal of producing a chronic implant that causes minimal nerve damage.

4. Investigate the possibility of extracting information about contact, grasped object slip, limb position and contact force from chronically recorded neural activity using the animal model and electrodes from parts 2 and 3.
   a. Devise recording, processing, and detection methods to extract this information from recorded neural activity in a restrained animal.
   b. Modify these methods as needed to function in an unrestrained animal and in the presence of stimulation artifacts associated with functional electrical stimulation.
   c. Record activity for periods of at least 6 months and devise functional measures to track any change in neural response over this time.
   d. Evaluate any histological changes in the nerves that occurred over the period of chronic recording and, if possible, correlate these changes to changes in functional response.

5. Cooperate with other investigators in the Neural Prosthesis Program by collaboration and sharing of experimental findings.

II. Summary of Progress in the Fourth Period

During the fourth reporting period we continued to study the 6 animals that were chronically implanted with either Multi-Contact Cuffs (MCCs) or Longitudinal Intra-Fascicular Electrodes (LIFEs). We periodically monitored compound action potentials, device impedances and trends in multi-channel recording selectivity using both electrical stimulation and mechanical stimulation of the digits under gas anesthesia. We also recorded multi-channel nerve and muscle activity during treadmill walking. As not all the trends had stabilized after the target monitoring period of 180 days, we extended the study of these animals for further periods.

III. Details of Progress in the Fourth Period

A. Multi-Channel Recordings from Peripheral Nerves

In our previous progress reports we described two basic multi-channel approaches that we are investigating for this contracted research, **multi-contact cuffs (MCCs)** and **longitudinal intra-fascicular electrodes (LIFEs).** In Yr 1 we implanted two arrays of either one type or the other type, in the left forelimb of each of three cats. A total of six cats are being studied for minimum periods of six months.

The results obtained from these experiments thus far were presented as a set of four papers at the IFESS '97/NP5 Meeting held at Simon Fraser University campus, Vancouver, on August 16-21, 1997. In this progress report we summarize these findings. We provide copies of the Proceedings papers in **Appendices 1-4.**

1. Properties of Multi-Contact Cuff (MCC) and Longitudinal Intra-Fascicular Electrode (LIFE) Arrays Implanted in Cat Forelimb Nerves

In the first year of this study we were able to determine that implantation of multiple electrodes of either type, MCC or LIFE, did not compromise the long-term stability of the cat median and ulnar nerves, based on the stability of their compound action potential (CAP) amplitudes. These results were communicated at the IFESS meeting by Hoffer et al., 1997 (see **Appendix 1**).

The characteristic CAP shapes that were recorded by MCC electrodes remained invariant from week to week. The average MCC CAP amplitudes closely

paralleled the fluctuations in the whole-nerve (tripolar cuff electrode) CAPs which occur typically and are attributed to connective tissue ingrowth and slow changes in the electrode impedance.

The CAPs recorded by LIFEs typically had more complex, variable shapes (suggesting electrode movement) and their amplitudes declined markedly in the first few weeks (suggesting encapsulation) but then started to stabilize and remained at acceptable levels.

These results suggest that both tested approaches to obtaining multiple sensory nerve signals from main nerve trunks are safe and may be clinically viable. With the designs that we used, the increased numbers of electrodes and lead wires per implanted nerve (up to 10 leads, compared to 2 leads for our cuffs with conventional tripolar electrodes) did not adversely affect either the nerves or the recorded signals.

2. Measurement of Selectivity

Methods for selective recording from nerves and muscles have existed for decades, but there is no widely accepted method for measuring their selectivity. Lichtenberg and De Luca [1] used multi-contact cuff (MCC) electrodes to record compound action potentials (CAPs) from peripheral nerves and found that normalized amplitudes of CAPs evoked by stimulation of different distal nerve branches had "striking" differences, such that the stimulated nerve branches were statistically distinguishable according to the CAPs. Sahin and Durand [2] did similar experiments and observed that the normalized amplitudes of the CAPs were not very different, but reached a similar conclusion that the selectivity of the recordings was statistically significant. Struijk and Haugland [3] defined a selectivity measure for the difference between CAPs evoked by stimulation of two nerve branches only.

It became clear to us that in order to evaluate alternative nerve recording methods, we first needed to develop a quantitative measure of selectivity which could be applied to all recording methods. Such a measurement should allow the evaluation and comparison of different methods as well as the quantitative assessment of improvements in selectivity attributable to electrode design modifications. Our newly developed selectivity measurement method that was communicated at the IFESS meeting by Chen et al., 1997 **(Appendix 2)** is a useful tool for evaluation and quantitative comparison of different multi-channel nerve recording methods.

3. Evaluation of Selectivity using Electrical Stimulation of Individual Digits

Progress Report #2 detailed our data collection and analysis protocols for experiments involving multi-channel selectivity with electrical stimulation of the digits. Each recording day produces a single overall "selectivity index" (SI) which is a relative metric of the effectiveness of the entire array of recording electrodes. We recorded a total of eight point-source ENG signals from either four pairs of LIFEs placed inside both the Median and Ulnar nerves, or from two MCCs with four bipolar electrode pairs each, placed around the Median and Ulnar nerves.

During the last reporting period we continued to compute the selectivity indices to determine if trends develop over time, given the trends in recording parameters of the electrodes shown by CAP recordings and by impedances.

As reported at the IFESS meeting by Strange et al., 1997 **(Appendix 3)** the maximum overall SI for MCCs was 61 (average = 44.9 ± 13.1, n = 30 recording days from three cats). The maximum SI for LIFEs was 80 (average = 67.3 ± 8.2, n = 22). In comparison, the maximum SI obtained by using the two single-channel tripolar cuffs placed on the two nerves was 48 (average = 38.7 ± 9.2, n = 48).

With repeated measurements over time, SIs calculated from the MCCs declined early on in one cat but remained relatively stable in the two other cats. The decline in SI in NIH 18 was likely a result of damage to the electrodes or lead wires in the Ulnar MCC and in the Ulnar tripolar cuff.

The 8-channel SIs in NIH 19 and 21 closely paralleled the tripolar cuff SIs over time, indicating that consistently high degrees of selectivity can be achieved over extended implant periods with multi-channel arrays of electrodes. LIFE SIs were generally higher than the SIs produced by MCCs and also remained relatively stable over the implant periods. This suggests that the CAP signal amplitude **ratios** remained largely unchanged even though the CAP signal amplitudes generally declined (Hoffer et al., 1997; **Appendix 1**).

The SIs from the tripolar cuffs in the three LIFE implants were the same as those from the last two MCC implants (SIs of approx. 40), indicating that the two types of multi-channel devices were tested in equivalent environments.

These results suggest that acquiring multi-dimensional sensory information from peripheral nerves with MCCs or LIFEs can lead to a higher degree of selectivity than achieved with single channel recording nerve cuffs on the same nerves.

4. Evaluation of Selectivity using Mechanical Stimulation of Individual Digits

The five-digit mechanical stimulator that was described in the previous progress report was used to produce two kinds of mechanical stimulation: slips along the long axis of each digit, or brief taps on each digit pad. The methodology and results to date were communicated by Christensen et al., 1997 **(Appendix 4).**

The selectivity values obtained for mechanical stimulation of the skin were lower overall than the values obtained with the same electrode arrays in response to electrical stimulation of the digits (Strange et al., 1997, **Appendix 3**). This is attributed to a larger standard deviation in the burst areas and a lower signal-to-noise observed with mechanical stimulation. Typical ratios of peak of filtered signal to background level of activity were about 3:1. Due to the asynchronous nature of neural bursts, the ENG amplitude was in the 5 - 10 microvolt range, whereas with electrical stimulation of digits the compound neural signal was about 10 times greater.

In spite of the small amplitudes of these neural signals, the selectivity of the multichannel arrays was impressive and it was possible to identify with high accuracy which digit was stimulated in each trial. Using the 8-channel LIFE arrays, the accuracy of digit identification in response to either kind of mechanical probing (slips or taps) of single digits ranged from 87% to 100% (mean: 97%, s.d.: 4%). Using the 8-channel MCCs, accuracy ranged from 71% to 92% (mean: 81%, s.d.: 8%). For comparison, the accuracy using a 2-channel array of tripolar cuffs on the same nerves ranged from 34% to 56% (mean: 44%, s.d.: 8%).

Although more advanced signal processing methods may lead to increased selectivity values and higher accuracy, we believe that the results obtained for digit identification are promising. Clearly, data collected with LIFEs had greatest selectivity and an average accuracy of 97%. The MCC recordings were not as selective, with average accuracy of 81%, but in comparison with using two tripolar recording cuffs on the median and ulnar nerves, the 8-channel MCC system also showed a dramatic increase in accuracy of digit identification. Also to be noted is that normal and slip perturbations could both be identified at the same rate with 8-channel systems. We conclude that multi-contact arrays of implanted nerve electrodes can be used to detect multi-dimensional sensory inputs applied to skin and should be suitable for control of multi-channel motor prostheses with FES.

B. Morphological Analysis of Cuffed Peripheral Nerves

During the fourth reporting period, D. Crouch completed a morphological study of the four cat forelimb median nerves that had been implanted with tripolar nerve cuffs for periods of 6-12 months during our previous contract period, for which the quality of the histological processing was good (cats NIH 12, 15, 16 and 17). These nerves were compared with their contralateral control nerves.

Axon numbers, axon diameters, fiber diameters and myelin thickness were measured. All myelinated axons >2μm in whole nerve cross-sections were counted. Axons in the core of the nerve and in a 50μm perimeter zone were separately assessed. These results were communicated by Crouch et al. at the 1997 IFESS meeting (see **Appendix 5**).

The average differences that were found between each cuffed Median nerve and its contralateral control nerve can be summarized as follows:

1. total number of myelinated axons per nerve	No significant difference
2. total nerve cross-sectional area (incl. connective tissue)	+29%
3. total intraaxonal cross-sectional area	- 16%
4. average axon diameter near nerve perimeter	- 10%
5. average myelinated fiber diameter near nerve perimeter	- 9%
6. average myelin thickness near nerve perimeter	- 8%
7. average axon diameter in nerve core	- 6%
8. average myelinated fiber diameter in nerve core	- 3%
9. average myelin thickness in nerve core	- 3%

The results indicate that the presence of nerve cuffs implanted around cat Median nerves for 6-12 month periods did affect the morphometry of the nerve axons, but only in subtle ways. There was no change in axon numbers. Cuffing caused a slight thinning of axons and in the thickness of their myelin, which was most prevalent for axons located near the outside perimeter of the nerve.

It is worth mentioning that all our implanted cats were group housed (not caged) and were free to jump up and down shelves up to 2 m high and climb on a vertical wire fence, activities which they all carried out vigorously and often on a daily basis. Once the cats had recovered from the implant surgery, at no time in the course of these experiments was any motor or sensory deficit apparent that could have been caused by the implanted cuffs.

On this basis, the successful long-term survival of both the implanted nerves and the multichannel electrode/multiwire lead assemblies bodes well for the anticipated safety of similar implants in humans.

IV. Plans for the Fifth Period

During the fifth reporting period, from October 1, 1997 to December 31, 1997, our objectives will consist of the following:

We will complete the study of CAPs and device impedances under anesthesia for the six animals currently under study. As mentioned, we decided to extend the periods of study beyond 180 days because some of the trends (e.g., the LIFE signal amplitudes) had not stabilized in some of the animals.

As each implanted animal completes its implant period, we will perform final acute experiments to retrieve devices and harvest nerve tissues for histological analysis. As well, in these acute experiments we will determine in the contralateral limbs which nerves will be best suited for recording proprioceptive signals during reaching movements.

We will instrument proprioceptive as well as primarily cutaneous nerves in the next series of implants. We will train these animals to perform forelimb reaching movements using the protocol communicated by Hansen et al., 1997 (**Appendix 6**). We will record multichannel ENGs as well as EMGs from six muscles during performance of reaching movements.

We will analyze data collected during walking on the treadmill and the multi-channel ENG data from the LIFEs and from the MCCs will be evaluated to determine if differences in the signals (and thus the selectivity) can be detected and to determine the reliability and utility of these signals.

We will continue the collaboration with Drs. Yoshida and Stein by implanting three more animals with LIFEs.

We will continue with the collaboration with Dr. Andrews by collecting data from the MCCs and virtual sensors during experiments under anesthesia and during walking on the treadmill.

V. Publications and Meetings

A. Publications Emerged from Contracts During the Fourth Period

The following papers report results from our currently contracted research:

1. Hoffer, J.A., K.D. Strange, P.R. Christensen, Y. Chen and K.Yoshida. Multichannel recordings from peripheral nerves: 1. Properties of multi-contact cuff (MCC) and longitudinal intra-fascicular electrode (LIFE) arrays implanted in cat forelimb nerves. IFESS/Neural Prostheses V Int'l. Conf., Vancouver, BC, pp. 239-240, 1997.

2. Chen, Y., P.R. Christensen, K.D. Strange and J.A. Hoffer. Multichannel recordings from peripheral nerves: 2. Measurement of selectivity. IFESS/Neural Prostheses V Int'l. Conf., Vancouver, BC, pp. 241-242, 1997.

3. Strange, K.D., P.R. Christensen, Y. Chen, K. Yoshida and J.A. Hoffer. Multichannel recordings from peripheral nerves: 3. Evaluation of selectivity using electrical stimulation of individual digits. IFESS/Neural Prostheses V Int'l. Conf., Vancouver, BC, pp. 243-244, 1997.

4. Christensen, P.R., Y. Chen, K. D. Strange and J. A. Hoffer. Multichannel recordings from peripheral nerves: 4. Evaluation of selectivity using mechanical stimulation of individual digits. IFESS/Neural Prostheses V Int'l. Conf., Vancouver, BC, pp. 217-218, 1997.

The following papers report results from both previous and currently contracted research:

5. Crouch, D., K.D. Strange and J.A. Hoffer. Morphometric analysis of cat median nerves after long-term implantation of nerve cuff recording electrodes. IFESS/Neural Prostheses V Int'l. Conf., Vancouver, BC, pp. 245-246, 1997.

6. Hansen, M., J.A. Hoffer, K.D. Strange and Y. Chen. Sensory feedback for control of reaching and grasping using functional electrical stimulation. IFESS/Neural Prostheses V Int'l. Conf., Vancouver, BC, pp. 253-254, 1997.

B. Meetings attended During the Fourth Period

The six papers listed above were presented at the joint IFESS/NP5 Meeting to be held in Vancouver, Canada on August 16-21, 1997.

VI. Appendices

Copies of the six papers listed above are attached as Appendices 1-6.

VII. References

[1] Haugland, M. and Hoffer, J.A. Slip information provided by nerve cuff signals: application in closed-loop control of functional electrical stimulation. IEEE trans. Rehab. Engng. 2:29-36, 1994.

[2] Lickel, A., Haugland, M.K., Haase, J. and Sinkjær, T. Restoration of lateral hand grasp using natural sensors. In Neuroprosthetics: from basic research to clinical applications, Springer, 1996, pp 327-343.

[3] Kallesøe, K., Hoffer, J.A., Strange, K. and Valenzuela, I. *Implantable Cuff having Improved Closure,* United States Patent No. 5,487,756 awarded January 30, 1996.

[4] B.K. Lichtenberg and C.J. De Luca: "Distinguishability of functionally distinct evoked neuroelectric signals on the surface of a nerve", IEEE T-BME, vol. 26, no. 4, 1979, pp.228-37.

[5] M. Sahin and D.M. Durand, "Selective recording with a multi-contact nerve cuff electrode," *Proc. 19th Ann. Int. Conf. of IEEE EMBS,* Amsterdam, 1996.

[6] J.J. Struijk, M.K. Haugland, and M. Thomsen, "Fascicle selective recording with a nerve cuff electrode," *Proc. 19th Ann. Int. Conf. of IEEE EMBS,* Amsterdam, 1996.

Multi-Channel Recordings from Peripheral Nerves: 1. Properties of Multi-Contact Cuff (MCC) and Longitudinal Intra-Fascicular Electrode (LIFE) Arrays Implanted in Cat Forelimb Nerves.

J.A. Hoffer (a,b), K.D. Strange (a), P.R. Christensen (b), Y. Chen (a) and K. Yoshida (c)

a. School of Kinesiology, Simon Fraser University, Burnaby, BC, V5A 1S6, Canada
b. School of Engineering Science, Simon Fraser University, Burnaby, BC, V5A 1S6, Canada
c. Division of Neuroscience, University of Alberta, Edmonton, AB, T6G 1K7, Canada.

ntroduction

ormal hand use relies heavily on sensory cues, especially tactile origin. After a spinal cord injury or stroke, the use paralyzed hand muscles may be restored with functional ctrical simulation (FES). Grip control is improved if sory feedback from either artificial or natural sensors can provided to the FES control unit. Slip-related sensory nals that are naturally generated by skin mechanoreceptors e been recorded from peripheral nerves using cuff ctrodes in animals [1] and humans [2] and can be ticularly useful for grip control with FES [1].

o adequately restore hand use with FES, it will be portant to obtain feedback signals that originate from ltiple sources: preferably from each fingertip, the palmar ger surfaces, and the palm.

Two basic recording approaches can be envisioned for aining neural signals of tactile origin from every digit. parate electrodes could be implanted in or on each ividual palmar digital nerve branch. Alternatively, several ctrodes could be implanted in or on the median and ulnar ves in the forearm, which together supply all five digits.

ne first approach was tested by placing a cuff electrode on almar digital nerve branch in two human volunteers and e encouraging results [2] but it is surgically demanding has high potential to cause nerve damage if applied to all palmar digital nerve branches. The second approach is nically simpler and the nerves are larger and less exposed nechanical damage, but it remains to be shown whether nals recorded from whole nerves can be sufficiently ctive to allow differentiation among the different digits.

this and the three following companion papers (Chen et Strange et al., Christensen et al.; this Meeting) we sent results of chronic experiments in which we eloped and evaluated two approaches for recording and lyzing multiple signals from larger nerve trunks in the limb: Multi-Contact Cuff (MCC) electrodes and arrays ongitudinal IntraFascicular Electrodes (LIFEs).

Methods

ne forelimb of the cat was used as model for the alyzed human forearm and hand. Electrical and mechanical stimulation of the individual digits under anesthesia was used to test the selectivity of multi-channel electrode arrays.

MCCs were constructed of silicone tubing and included an interlocking cuff closing mechanism [3]. Four pairs of fine wire recording electrodes were placed inside each cuff (the design is described in a patent application). In other cats, sets of four pairs of LIFEs (see description by Yoshida et al. this Meeting) were sewn into the left ulnar and median nerves. Conventional tripolar recording cuff electrodes and bipolar stimulating cuff electrodes were implanted distal and proximal to the MCCs or LIFEs (Fig. 1). The cuffs were used in this study to periodically monitor compound action potential (CAP) shape and amplitude under anesthesia to assess the status of nerves and electrodes over six months.

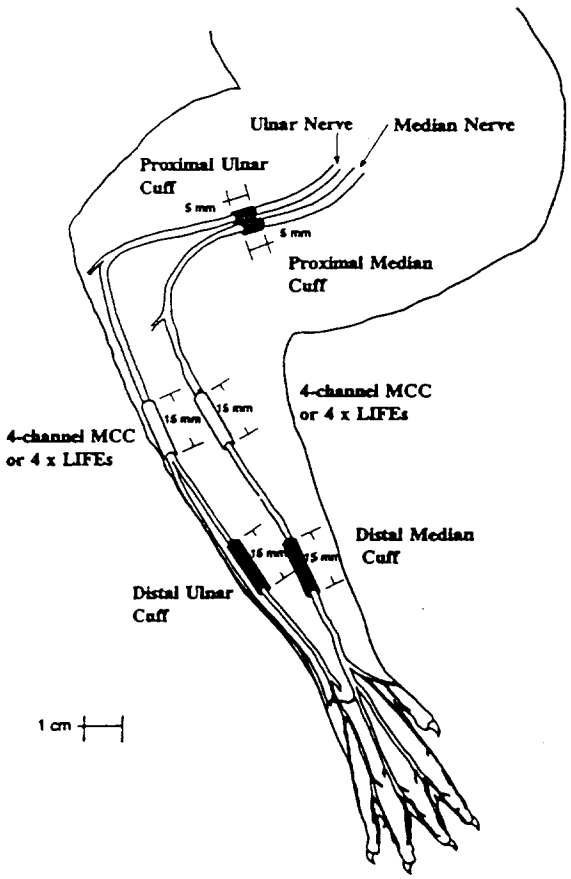

Fig. 1: Schematic diagram of left cat forelimb and implanted devices. Multi-electrode (MCC or LIFE) arrays were installed between proximal stimulating and distal recording cuffs.

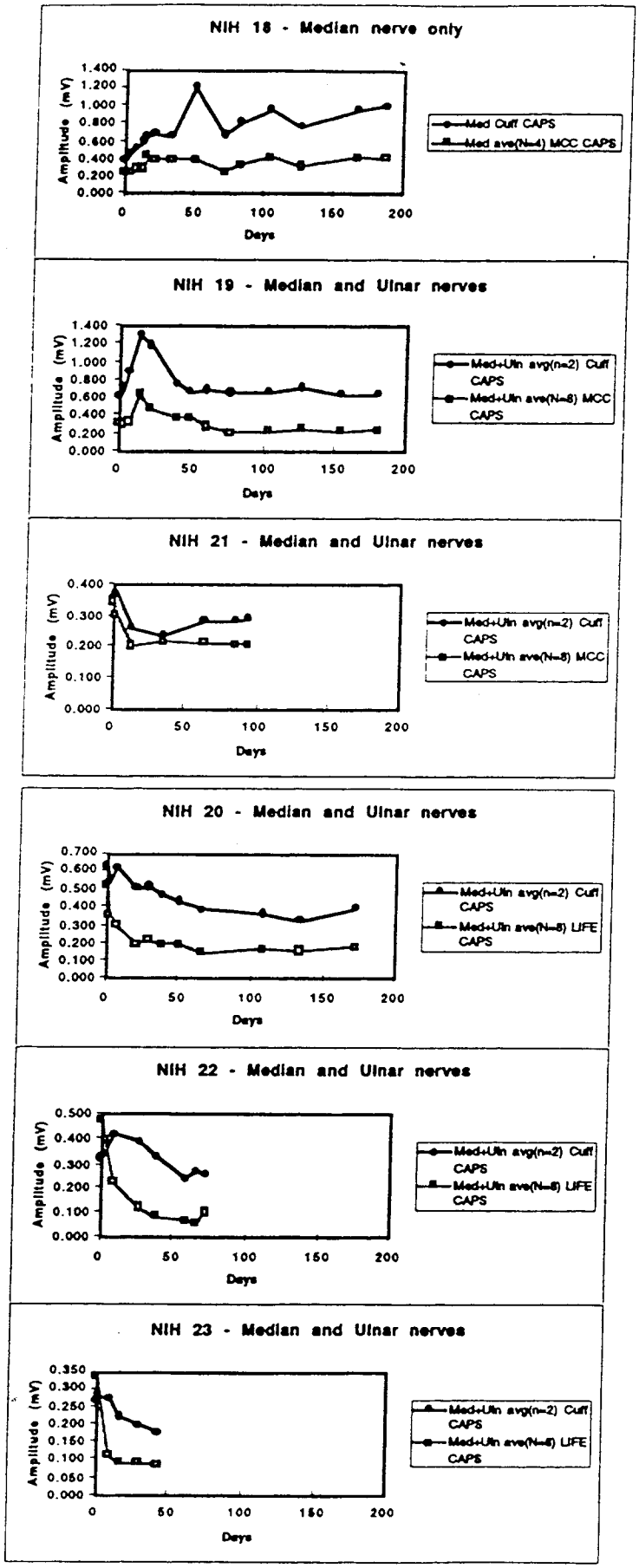

Fig. 2. Time course of nerve CAP amplitudes in six cats implanted with MCC (top) or LIFE (bottom) electrode sets.

III. Results

There are two main findings from this study. First, we were able to determine that implantation of multiple electrodes of either type, MCC or LIFE, did not compromise the long-term stability of the median and ulnar nerves based on the stability of their CAP amplitudes. Second, with only some exceptions the selective electrode pairs continued to function and provided good signals over the duration of the study. Several electrode lead wires (ulnar nerve MCC and recording cuff) broke soon after implantation in NIH 18 and about five LIFE wires broke during the period of this study.

Characteristic CAP shapes were recorded by MCC electrode pairs and remained invariant from week to week. The MCC CAP amplitudes (average CAPs from up to eight electrode pairs per animal are shown in Fig. 2, light traces) closely paralleled the typical fluctuations in the whole-nerve CAPs (Fig. 2, dark traces) which were attributed to connective tissue ingrowth and changes in the electrode impedance measured at 1kHz. The CAPs recorded by LIFEs typically had more complex, variable shapes (suggesting electrode movement) and their amplitudes declined markedly in the first few weeks (suggesting encapsulation) but then started to stabilize and remained at acceptable levels.

IV. Discussion and Conclusions

These results suggest that both tested approaches to obtaining multiple sensory nerve signals from main nerve trunks were safe and may be clinically viable. With the designs that were used, the increase in numbers of electrodes and lead wires per implanted nerve did not adversely affect either the nerves or the recorded signals.

References

[1] Haugland, M. and Hoffer, J.A. Slip information provided by nerve cuff signals: application in closed-loop control of functional electrical stimulation. IEEE trans. Rehab. Engng. 2:29-36, 1994.

[2] Lickel, A., Haugland, M.K., Haase, J. and Sinkjær, T. Restoration of lateral hand grasp using natural sensors. In Neuroprosthetics: from basic research to clinical applications, Springer, 1996, pp 327-343.

[3] Kallesøe, K., Hoffer, J.A., Strange, K. and Valenzuela, I. *Implantable Cuff having Improved Closure,* United States Patent No. 5,487,756 awarded January 30, 1996.

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APPENDIX 2

Multi-Channel Recordings from Peripheral Nerves: 2. Measurement of Selectivity

Y. Chen (a), P.R. Christensen (b), K.D. Strange (a), and J.A. Hoffer (a,b)

a. School of Kinesiology, Simon Fraser University, Burnaby, BC, V5A 1S6, Canada
b. School of Engineering Science, Simon Fraser University, Burnaby, BC, V5A 1S6, Canada

I. Introduction

Methods for selective recording from nerves and muscles have existed for decades, but there is no widely accepted method for measuring their selectivity. Lichtenberg and De Luca [1] used multi-contact cuff (MCC) electrodes to record compound action potentials (CAPs) from peripheral nerves and found that normalized amplitudes of CAPs evoked by stimulation of different distal nerve branches had "striking" differences, such that the stimulated nerve branches were statistically distinguishable according to the CAPs. Sahin and Durand [2] did similar experiments and observed that the normalized amplitudes of the CAPs were not very different, but reached a similar conclusion that the selectivity of the recordings was statistically significant. Struijk and Haugland [3] defined a selectivity measure for the difference between CAPs evoked by stimulation of two nerve branches only.

It became clear to us that in order to evaluate alternative nerve recording methods, we first needed to develop a quantitative measure of selectivity which could be applied to all recording methods. Such a measurement should allow the evaluation and comparison of different methods as well as the quantitative assessment of improvements in selectivity attributable to electrode design modifications.

II. Methods

Theoretical Development

Although the selectivity of a nerve recording method must ultimately be tested in applications for which the recording method is designed, measurements of selectivity are better taken under controlled conditions. For this reason, we developed a testing method based on recording CAPs evoked by electrical stimulation of nerve branches.

We let $V_i = (v_{i1}, v_{i2}, \ldots, v_{iN})$, for $i = 1, 2, \ldots, M$, be a vector whose N elements are features derived from the CAPs recorded from a nerve trunk with a multi-channel electrode array when the *ith* nerve branch of M branches is stimulated. The feature elements can be amplitudes, areas, etc., of the CAP waveforms. A feature space is formed with the M feature vectors for the M nerve branches. First, we normalize all feature vectors to unitary magnitude. Thus, the normalized feature vector is

$$W_i = (w_{i1}, w_{i2}, \ldots, w_{iN}) = \frac{V_i}{|V_i|}$$

where $|V_i| = \sqrt{v_{i1}^2 + v_{i2}^2 + \ldots + v_{iN}^2}$. We then define the Euler distance between two normalized feature vectors as their difference. Further, the distance is measured as percentage of the maximum possible distance (which can be shown to be $\sqrt{2}$). That is, the distance between feature vectors for the *ith* and *jth* nerve branches is:

$$d_{ij} = \frac{100}{\sqrt{2}} \sqrt{(w_{i1} - w_{j1})^2 + (w_{i2} - w_{j2})^2 + \ldots + (w_{iN} - w_{jN})^2}$$

Such distances range from 0 to 100 (as percent of the maximum distance).

We define the Selectivity Index of the recording method for the *kth* nerve branch, S_k, as the average distance between the *kth* feature vector and all the other feature vectors:

$$S_k = \frac{1}{M-1} \sum_{j=1, j \neq k}^{M} d_{kj}.$$

Finally, we define the Overall Selectivity Index, S, of the recording method as the average of the individual selectivity indices for all the stimulated nerve branches:

$$S = \frac{1}{M} \sum_{k=1}^{M} S_k$$

Since the selectivity indices S_k and S are averages of certain distance values which range from 0 to 100, S_k and S also range from 0 to 100, with 0 meaning no selectivity at all and 100 the maximum theoretically achievable selectivity.

Experiments

In three acute experiments under anesthesia, 4-channel multi-contact cuff (MCC) electrodes of two designs (conventional and improved, each containing four electrode pairs) were placed around cat sciatic nerves in the mid-thigh

region. In each of the experiments, five to eight nerve branches from the following list were dissected free: the common peroneal (CP), deep and superficial peroneal (DP and SP), Tibial (Tib), first and second parts of Tibial (Tib1 and Tib2), lateral gastrocnemius-soleus (LGS), medial gastrocnemius (MG), sural, perforant branch of biceps (Perf), and plantaris (PL). Sets of CAPs were recorded in response to stimulating each of the isolated sciatic branches and the peak-to-peak amplitudes of CAPs in each data set were used to form a feature vector.

In the first experiment (Acute #1), an improved MCC (I-MCC) electrode of proprietary design was tested with five nerve branches being stimulated. In the second experiment (Acute #2), an I-MCC and a conventional MCC (C-MCC) electrode were tested in sequence using the same testing protocol. Eight nerve branches were stimulated. In the third experiment (Acute #3), an I-MCC and a C-MCC electrode were tested in alternated sequence with the first and third trials testing the C-MCC (C-1 and C-2) and the second and forth trials testing the I-MCC (I-1 and I-2). Six nerve branches were stimulated in each of the trials.

III. Results

Selectivity indices were calculated according to the definition given in this report for all three acute experiments. The results are presented in Fig. 1, 2, and 3, respectively. The average selectivity indices ranged between 26 and 38 for the improved MCC electrodes, and between 11 and 14 for the conventional MCC electrodes.

IV. Discussion

As can be seen from Fig. 1, 2, and 3, the improved MCC electrodes always provided better selectivity than the conventional MCC electrodes, regardless of experimental subject or sequence of trials. The selectivity measurements were also repeatable, as demonstrated in Acute #3.

V. Conclusion

The newly developed selectivity measurement method that is presented in this report is a useful tool for evaluation and quantitative comparison of different multi-channel nerve recording methods.

References

[1] B.K. Lichtenberg and C.J. De Luca: "Distinguishability of functionally distinct evoked neuroelectric signals on the surface of a nerve", IEEE T-BME, vol. 26, no. 4, 1979, pp.228-37.

[2] M. Sahin and D.M. Durand, "Selective recording with a multi-contact nerve cuff electrode," *Proc. 19th Ann. Int. Conf. of IEEE EMBS*, Amsterdam, 1996.

[3] J.J. Struijk, M.K. Haugland, and M. Thomsen, "Fascicle selective recording with a nerve cuff electrode," *Proc. 19th Ann. Int. Conf. of IEEE EMBS*, Amsterdam, 1996.

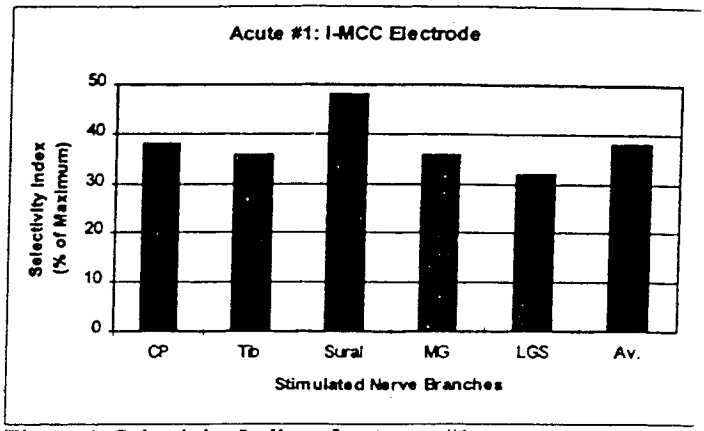

Figure 1. Selectivity Indices for Acute #1

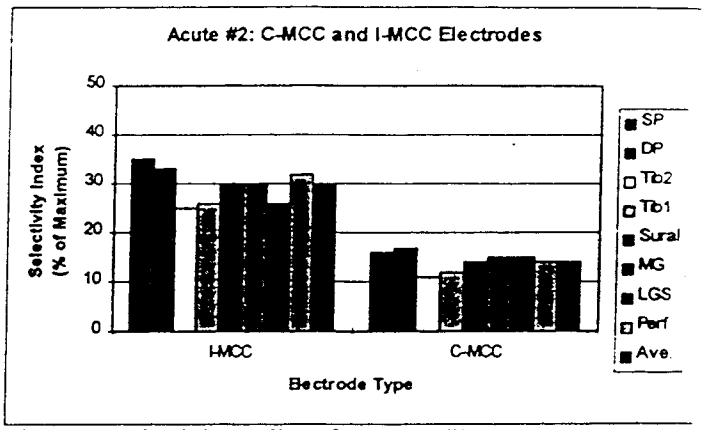

Figure 2. Selectivity Indices for Acute #2

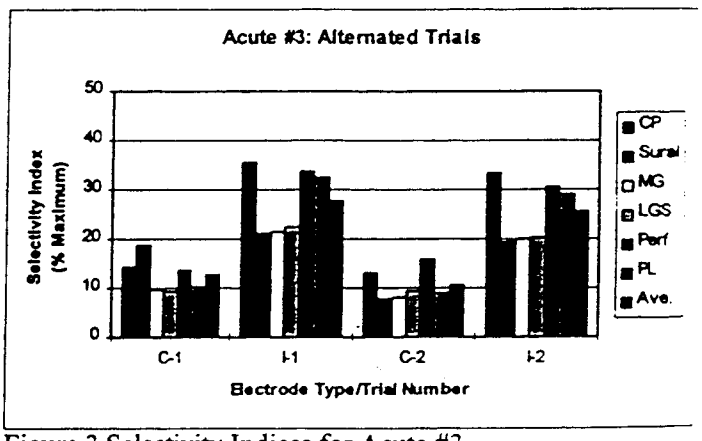

Figure 3 Selectivity Indices for Acute #3.

Acknowledgments

The authors thank Tiffany Blasak and Josh Lawson for excellent technical support. Funding was provided by NIH Contract No. NIH-NINDS-NO1-NS-6-2339 and the Canadian NeuroScience Network of Centres of Excellence, Theme 6, Project A3.

APPENDIX 3

Multi-Channel Recordings from Peripheral Nerves: 3. Evaluation of Selectivity using Electrical Stimulation of Individual Digits.

K.D. Strange (a), P.R. Christensen (b), Y. Chen (a), K. Yoshida (c), and J.A. Hoffer (a,b)

a. School of Kinesiology, Simon Fraser University, Burnaby, BC, V5A 1S6, Canada
b. School of Engineering Science, Simon Fraser University, Burnaby, BC, V5A 1S6, Canada
c. Division of Neuroscience, University of Alberta, Edmonton, AB, T6G 1K7, Canada.

I. Introduction

We present here selectivity measurements for two types of permanently implanted nerve recording electrode arrays, Multi-Contact Cuff (MCC) electrodes and Longitudinal Intra-Fascicular Electrodes (LIFEs). The overall goal of these studies is to investigate the feasibility and utility of developing selective multi-channel ENG recording methods that can provide sensory feedback for FES systems.

II. Methods

The left Median and Ulnar nerves of three cats were each instrumented with 4-channel MCCs (each having four pairs of small electrodes placed externally to the nerves). The same nerves in three other cats were each instrumented with four pairs of LIFEs (implanted in the nerves), with devices located as in Fig. 1.

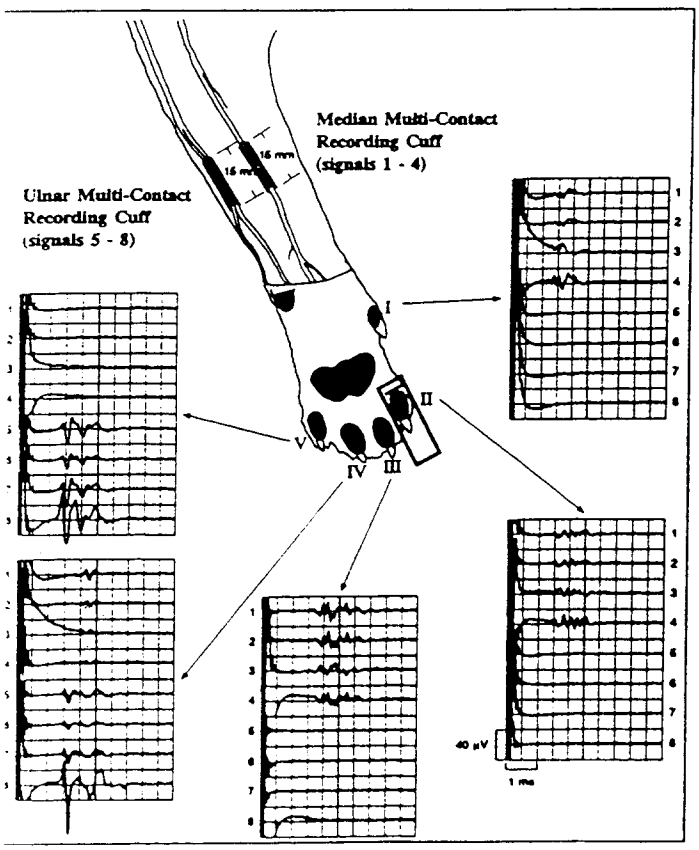

Fig. 1: Electrical stimulation of the digits (NIH 19, day 0)

Conventional nerve stimulating cuffs and single-channel tripolar recording cuffs were implanted proximal and distal to the multi-channel devices, respectively, to evaluate the status of the whole nerve over periods of at least six months (Hoffer et al., this conference).

In experiments under anesthesia repeated every 1-4 weeks, the nerve branches supplying each of the five digits were transcutaneously stimulated with external circumferential electrodes within a cuff that was placed around the digit, using monophasic 10 mA, 50μs pulses, similar to acute experiments with direct nerve stimulation [1,2], and five sets of eight channels of evoked, synchronous ENG were collected. Figure 1 shows a representative set of evoked ENG data recorded with MCCs during stimulation of each digit.

The peak-to-peak amplitude of each evoked ENG compound action potential (CAP) was determined from averages of ten trials. The sets of CAP amplitudes were analyzed by applying the Euler distance method (Chen et al., this conference), and computing the average distance (or Selectivity Index, SI) between the vectors for each pair of digits. The maximum theoretical SI using this method is 100.

III. Results

Each recording day led to a single SI value for that experiment. Each cat was monitored over the duration of the implant with the longest lasting more than 180 days. Figures 2 and 3 present the trends in SIs over time for the three animals implanted with MCCs and the three animals implanted with LIFEs, respectively.

The maximum overall SI for MCCs was 61 (average = 44.9 ± 13.1, n = 30 recording days from three cats). The maximum SI for LIFEs was 80 (average = 67.3 ± 8.2, n = 22). In comparison, the maximum SI obtained by using the two single-channel tripolar cuffs placed on the two nerves was 48 (average = 38.7 ± 9.2, n = 48).

With repeated measurements over time, SIs calculated from the MCCs declined early on in one cat (Fig. 2, NIH 18, top panel) but remained relatively stable in the two other cats.

We suspect that the decline in SI in NIH 18 was a result of damage to the electrodes or lead wires in the Ulnar MCC and in the Ulnar tripolar cuff.

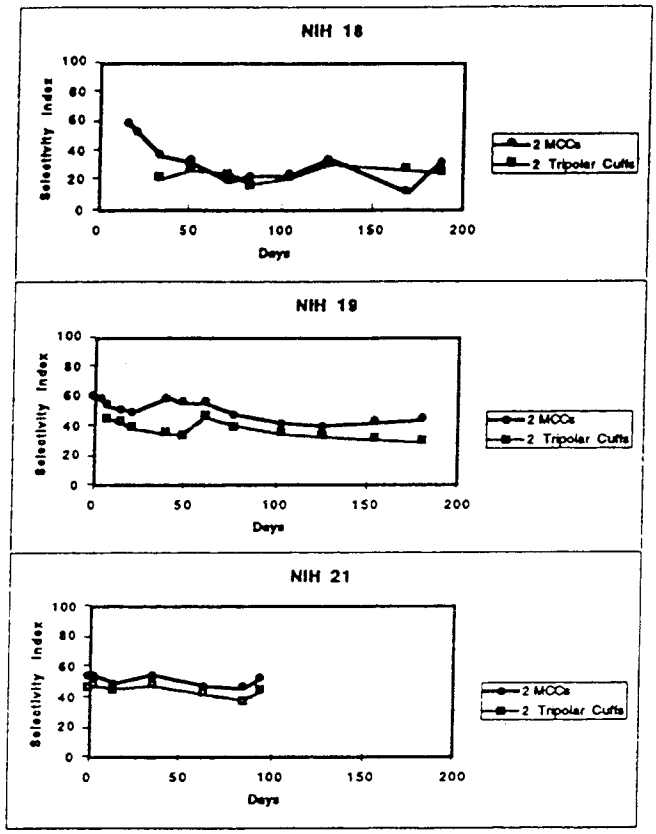

Fig. 2: Trends in Selectivity with MCCs

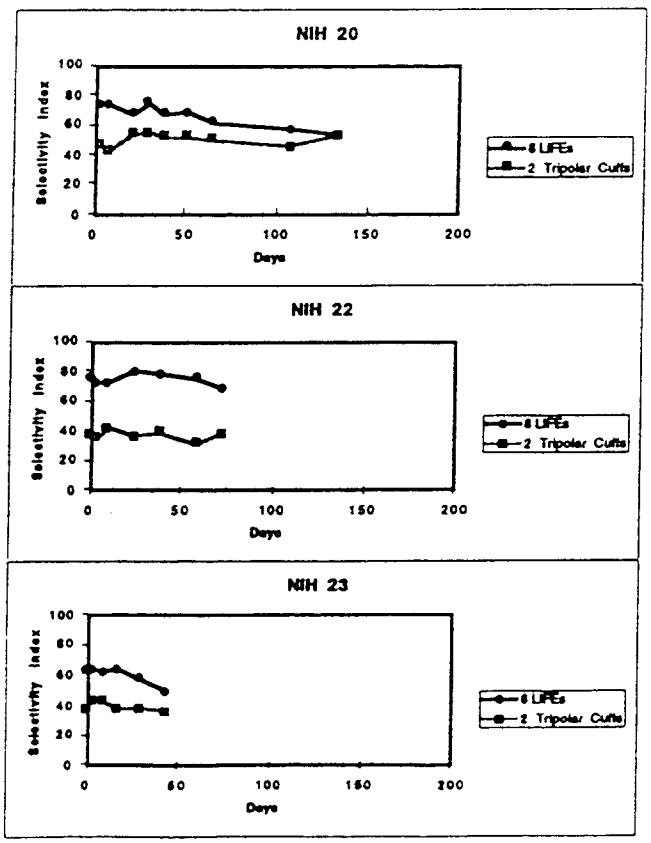

Fig. 3: Trends in Selectivity with LIFEs

The 8-channel SIs in NIH 19 and 21 closely paralleled the tripolar cuff SIs over time, indicating that consistently high degrees of selectivity can be achieved over extended implant periods with multi-channel arrays of electrodes. LIFE SIs were generally higher than the SIs produced by MCCs and also remained relatively stable over the implant periods.

The SIs from the tripolar cuffs in the three LIFE implants were the same as those from the last two MCC implants (SIs of approx. 40), indicating that the two types of multi-channel devices were tested in equivalent environments.

IV. Discussion

LIFEs implanted in nerves provide a more intimate interface with sensory nerve fibres than do external MCC electrodes. and thus provide much more selective signals in response to digit stimulation. Each LIFE pair produces a nearly binary signal depending if the surrounding nerve fibres innervate the stimulated digit. Normally, only one or two LIFE pairs in an array of four in the nerve showed an evoked signal. The four external electrode pairs in each MCC tended to show similar patterns of evoked neural activity, with differences in amplitude that presumably depended on electrode proximity to the stimulated fibres.

The longevity and stability of the individual neural recording devices may play a role in the stability of the SIs, although we have found that trends in the evoked neural compound action potentials (Hoffer et al., this meeting) do not necessarily affect the SI over time, as shown in Figs. 2 and 3.

V. Conclusion

These results suggest that acquiring multi-dimensional sensory information from peripheral nerves with MCCs or LIFEs can lead to a higher degree of selectivity than achieved with single channel recording nerve cuffs on multiple nerves. Multi-channel state feedback is expected to be useful for FES applications that require multi-digit manipulations.

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Acknowledgments

The authors thank Tiffany Blasak and Josh Lawson for excellent technical support. Funding was provided by NIH Contract No. NIH-NINDS-NO1-NS-6-2339, and the Canadian Neuroscience Network of Centres of Excellence, Theme 6, Project A3.

APPENDIX 4

Multi-Channel Recordings from Peripheral Nerves: 4. Evaluation of Selectivity using Mechanical Stimulation of Individual Digits

P. R. Christensen (b), Y. Chen (a), K. D. Strange (a), K. Yoshida (c), and J. A. Hoffer (a,b)

a. School of Kinesiology, Simon Fraser University, Burnaby, BC, V5A 1S6, Canada
b. School of Engineering Science, Simon Fraser University, Burnaby, BC, V5A 1S6, Canada
c. Division of Neuroscience, University of Alberta, Edmonton, AB, T6G 1K7, Canada

Introduction

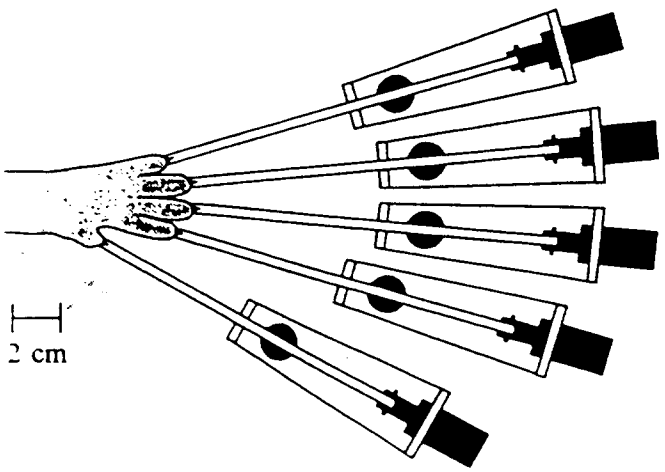
ɔmpanion studies (Chen et al., Strange et al., this meeting) ·e focused on determining nerve recording selectivity using ectrical stimulation. For this study we used the same cats, ;uipped with the same chronically implanted Multi- ɔntact Cuff (MCC) or Longitudinal IntraFascicular ectrode (LIFE) recording arrays, to investigate the lectivity of recordings in response to mechanical mulation of the skin of individual digit pads.

· using mechanical inputs, a step was taken into the "real ɔrld" because the measured nerve activity was generated mechanoreceptors that are normally stimulated by skin ntact or slip.

Methods

a model of a paralyzed human forearm, we used the left ·earm of chronically implanted, anesthetized cats. A mputer-controlled 5-digit manipulator (see Fig. 1) nerated two types of mechanical perturbation: brief lentations normal to a digit pad lasting 50 ms or fast slips ng that pad, delivered up to twice/s. We recorded eight annels of electroneurographic (ENG) activity from dian and ulnar nerves with MCC and LIFE arrays, and ɔ whole-nerve ENG channels with conventional tripolar ording cuffs (described by Hoffer et al., this meeting).

e ENG data were later digitized, rectified, and filtered to ain smoothed envelopes of the neural activity that



g 1: Five-digit manipulator positioned under cat forepaw

occurred in response to mechanical stimulation of the five digits with both stimulus types. Features of the envelopes were used to form sets of feature vectors from which selectivity measurements were made (Chen et al., this meeting). Figure 2 shows seven channels of filtered neural responses while stimuli were sequentially applied to digits 3, 4, and 5. Shaded regions correspond to data sections used to measure ENG area features. (The 4th recording channel from the ulnar nerve could not be used because large amplitude noise contaminated the nerve signal.)

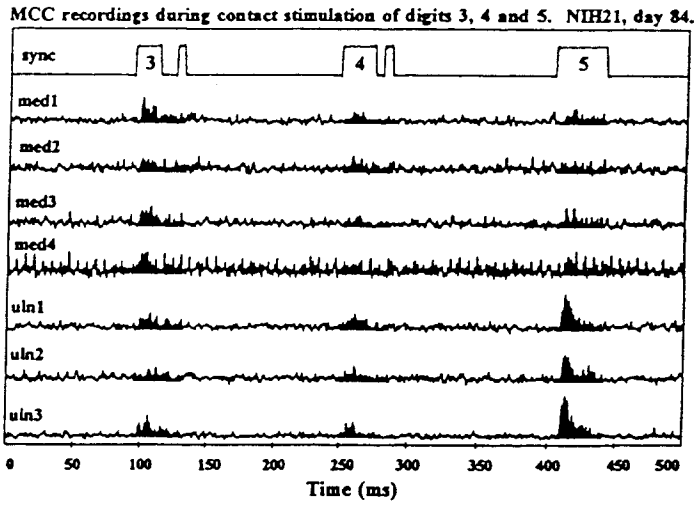


Fig. 2: Example of filtered neural signals

To identify which digit was stimulated in each trial, a discriminant analysis classified the data into five groups based on the multi-channel feature data. The accuracy of digit identification was determined by comparing the predicted to the known stimulated digit. Figure 3 shows the "clouds" of data that resulted from the analysis. Function 1 and Function 2 refer to the first two of four canonical functions that were used to classify the data. (The other two canonical functions that were necessary to separate the data into five categories could not be shown in this 2-dimensional scatter plot.)

III. Results

Table 1 shows results of selectivity analyses and accuracy tests after mechanical stimulation of the five digits. Data were collected for both 8-channel (i.e., MCC or LIFE) and 2-channel (i.e., standard cuff) electrode arrays. Area

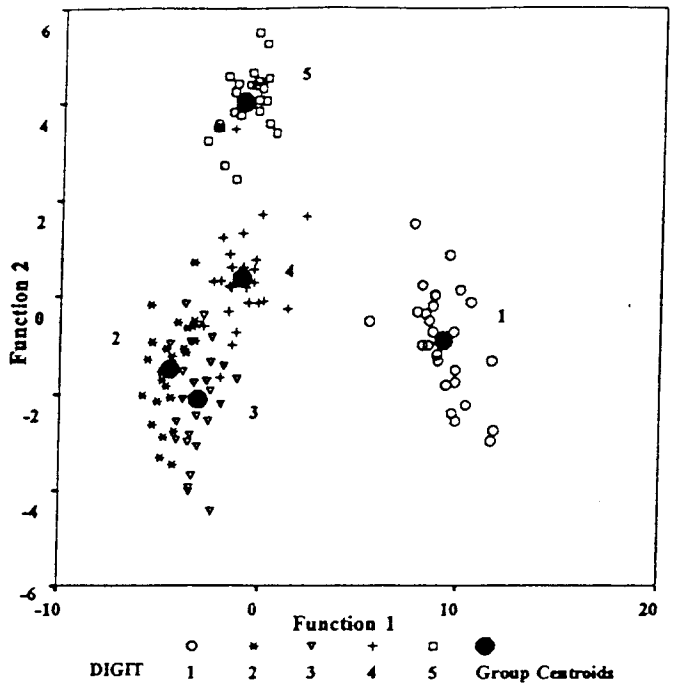


Fig. 3: Scatter plot of area feature data for NIH19, day 180, slip perturbations. Correct digit identification: 92%

features of the filtered neural bursts were used to derive these values.

The accuracy of digit identification using 8-channel LIFE arrays ranged from 87% to 100% (mean: 97%, s.d.: 4%). Using the 8-channel MCCs, accuracy ranged from 71% to 92% (mean: 81%, s.d.: 8%). Accuracy using 2-channel

Table 1: Selectivity index and accuracy of identification of which of five digits was stimulated using 8- and 2-channel nerve recording arrays
(S = slip, N = normal stimulation.)

Subject	Day	8-channel selectivity index (%)	8-channel accuracy (%)	2-channel selectivity index (%)	2-channel accuracy (%)
NIH19	154-S	16	76		
(MCC)	180-N	14	82		
	180-S	20	92		
NIH21	84-N	14	85	6	44
(MCC)	84-S	6	71	4	35
	94-N	7	77	7	40
	94-S	9	84	5	37
	99-N	13	89	7	34
	99-S	7	71	5	38
NIH22	24-S	23	99	4	34
(LIFE)	58-N	25	100	6	50
	58-S	28	99	10	54
	65-N	25	87	12	56
	65-S	28	98	8	50
	72-N	33	96	10	52
	72-S	32	98	6	41
NIH23	29-N	18	99	10	53
(LIFE)	29-S	18	98	8	47
	43-N	28	96	11	56
	43-S	17	94	10	35

arrays of two tripolar cuffs ranged from 34% to 56% (mea 44%, s.d.: 8%).

A correlation coefficient of 0.79 resulted between selectivit index and percentage accuracy using 8-channel systems.

IV. Discussion

The selectivity values obtained for mechanical stimulatio of the skin were lower overall than the values obtained wit the same electrode arrays in response to electrica stimulation of the digits (Strange et al., this meeting). Thi is attributed to a larger standard deviation in the burst area and a lower signal-to-noise observed with mechanica stimulation. Typical ratios of peak of filtered signal t background level of activity were about 3:1. Due to th asynchronous nature of neural bursts, the ENG amplitud was in the 5 - 10 microvolt range, whereas with electrica stimulation of digits the compound neural signal was abou 10 times greater.

Although more advanced signal processing methods ma lead to increased selectivity values and higher accuracy, w believe that the results obtained for digit identification ar promising. Clearly, data collected with LIFEs had greates selectivity and an average accuracy of 97%. The MCC recordings were not as selective, with average accuracy o 81%, but in comparison with using two tripolar recordin cuffs on the median and ulnar nerves, the 8-channel MCC system also showed a dramatic increase in accuracy of digi identification. Also to be noted is that normal and slip perturbations could both be identified at the same rate wit 8-channel systems.

The moderate correlation that exists between selectivity index and percentage accuracy for the 8-channel array gives a useful benchmark for determining the level o selectivity that will be needed to obtain a desired level o accuracy in more general applications.

V. Conclusion

Multi-contact arrays of implanted nerve electrodes can be used to detect multi-dimensional sensory inputs applied t skin and should be suitable for control of multi-channe motor prostheses with FES.

Acknowledgments

We thank Alex Szolnoki of SFU's Science Technical Centre for construction of the digit manipulators, and Tiffany Blasak and Josh Lawson for excellent care of the animals. Funding was provided by NIH Contract No. NIH-NINDS-NO1-NS-6-2339 and the Canadian Neuroscience Network of Centres of Excellence, Theme 6, Project A3.

APPENDIX 5

Morphometric Analysis of Cat Median Nerves After Long-Term Implantation of Nerve Cuff Recording Electrodes

D.A. Crouch, K.D. Strange and J.A. Hoffer

School of Kinesiology, Simon Fraser University, Burnaby, BC, V5A 1S6, Canada

I. Introduction

This study aimed to quantify the effects on axons of chronic implantation of tripolar nerve cuff recording electrodes. Since there are no statistically sound methods for sampling whole nerve cross sections [1], this study measured all the axons (>2μm) from all cross sections studied. Four cat median nerves implanted for 6-12 months were compared to unoperated contralateral control nerves in the same animal. The morphologies of axons located near the nerve perimeter were compared to axons nearer to the interior.

II. Methods

Fresh median nerve samples were fixed and embedded as per Dyck et al [2] and Bancroft and Stevens [3]. Samples were dehydrated, osmicated and then embedded in Jembed 812. Sections cut with a glass knife were counterstained with a 2:1 mixture of Richardson's stain [4] and Toluidine Blue. Cross sections were magnified under oil immersion, digitized and analyzed using an image analysis software package (Optimas Corp, Seattle, WA). Optimas allowed for semiautomatic analysis of the outlines of axons but allowed operator intervention when necessary.

Axon diameter, fiber (axon plus myelin) diameter and myelin thickness were used as morphometric indicators of nerve integrity. Diameters were calculated from the measured perimeters of the axons and the fibers respectively under the assumption that the axons were perfect circles *in vivo*. Myelin thickness was calculated as half the difference between axon diameter and fiber diameter. Conduction area was calculated as the total number of axons multiplied by the average axon area (in mm^2). Total nerve area was the sum of all connective tissue and conduction area (in mm^2).

Axons located in the outer 50 μm "shell" of the nerve cross section were binned separately from the remainder of the cross section to investigate whether selective damage occurred in this zone. These axons are hereafter referred to as perimeter axons.

III. Results

A general thickening of the connective tissue layer was observed around the implanted nerves. As evidenced by the increase in the total nerve cross sectional area (Table 1). This change is a natural outcome of the protective immune response to the introduction of a foreign body and has been well documented in the past.

Table 1 also demonstrates a less dramatic diminution in the amount of nerve area devoted to signal conduction. The significance of these changes with respect to signal conduction have yet to be elucidated.

		Control	Cuffed	% control
N12	Nerve Area	0.630	0.821	130
	Conduction Area	0.104	0.092	89
N15	Nerve Area	0.503	0.667	134
	Conduction Area	0.111	0.079	72
N16	Nerve Area	0.485	0.536	113
	Conduction Area	0.113	0.084	74
N17	Nerve Area	0.434	0.628	143
	Conduction Area	0.104	0.106	102
Overall	Nerve Area	0.513	0.665	129
	Conduction Area	0.108	0.090	84

Table 1: Summary table of conduction area (in mm^2) and total nerve area (in mm^2).

Table 2 summarizes the changes that occurred in three key neural measures for axons located in the outer 50μm shell of the nerve cross section. There was a general, statistically significant trend toward reduced axon diameters and reduced fiber diameters in the cuffed nerves. The same results were observed for myelin thickness.

The situation observed in the nerve perimeter tended to be similar to the interior of the nerve (Table 3). Axon diameter, fiber diameters and myelin thickness were significantly reduced in the cuffed nerves.

A comparison of the relative percentage changes apparent in perimeter axons as compared to interior axons suggests that perimeter axons experienced a greater change in axon diameter, fiber diameter and myelin thickness than did interior axons. This result is consistent with previous nerve compression studies [5,6].

Perimeter Cells		Control	Cuffed	% control
N12	# axons	627	486	77.5
	axon dia	6.3±3.0	6.3±3.3	100.6
	fiber dia	10.4±4.3	10.1±4.5	96.4
	myelin	2.1±0.9	1.9±0.9	89.9 **
N15	# axons	413	563	136.3
	axon dia	6.8±3.5	5.8±2.7	86.0 **
	fiber dia	11.9±5.3	10.0±4.3	84.1 **
	myelin	2.6±1.2	2.0±1.1	81.6 **
N16	# axons	567	433	76.37
	axon dia	6.8±2.9	6.1±2.7	88.9 **
	fiber dia	11.3±4.3	9.8±3.9	87.1 **
	myelin	2.2±0.9	1.9±0.8	84.4 **
N17	# axons	797	718	90.1
	axon dia	6.6±2.6	5.8±2.4	86.6 **
	fiber dia	10.3±3.5	9.9±3.8	96.12
	myelin	1.8±0.8	2.1±0.9	113.1 **

** Significant at p= 0.001

Table 2: Summary of relative changes in axon diameter, fiber diameter and myelin thickness for perimeter axons in control and cuffed median nerves in 4 cats.

Inner Cells		Control	Cuffed	% control
N12	# axons	4892	4976	101.7
	Axon Dia	6.3±2.7	6.3±2.8	101.9
	Fiber Dia	10.4±4.2	10.3±4.1	99.3
	Myelin	2.1±0.9	2.0±0.9	95.6 **
N15	# axons	4189	4476	106.85
	Axon Dia	6.8±2.9	5.6±2.4	83.1 **
	Fiber Dia	12.2±4.5	10.5±4.1	86.3 **
	Myelin	2.7±1.3	2.4±1.2	90.7 **
N16	# axons	4699	3810	81.1
	Axon Dia	6.4±2.7	6.2±2.7	95.8 **
	Fiber Dia	11.1±4.1	10.2±4.2	91.9 **
	Myelin	2.3±1.0	2.0±1.0	86.6 **
N17	# axons	3849	3992	103.7
	Axon Dia	6.5±2.6	6.2±2.6	94.6 **
	Fiber Dia	10.4±3.6	10.6±4.0	102.0
	Myelin	1.92±0.8	2.2±0.9	114.6 **

** Significant at p= 0.001

Table 3: Summary of relative changes in axon diameter, fiber diameter and myelin thickness for interior axons in control and cuffed median nerves in 4 cats.

IV. Discussion

This study has demonstrated that circumferential nerve cuffs can cause small but statistically significant changes in three important morphometric measures. Axon diameters, fiber diameters and myelin thickness were decreased in both the perimeter and interior zones. Greater effect was apparent in the perimeter.

This study also confirmed that there are no suitable sampling techniques for morphometric nerve analysis. Wide interfasicular variations of all the morphometric measures were observed in most cases. If morphometric techniques are to be used as a measure of nerve health, then it is essential that the dimensions of all the axons in the entire cross section be used. Large sampling errors can result if this guideline is not followed.

V. Conclusion

Small but statistically significant changes in axon diameter, fiber diameter and myelin thickness were observed following the log-term implantation of circumferential nerve cuff electrodes on cat median nerves. Decreases in axon diameter, fiber diameter and myelin thickness were greater in more peripherally located axons than in axons located closer to the center of the nerve. While the morphometric data are clear, the neurophysiological significance of these changes has yet to be elucidated. It is unclear, for instance, whether a statistically significant 0.3 μm reduction in average axon diameter is functionally significant to the nerve as a whole. These issues will be examined in the near future.

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Acknowledgments

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Sensory Feedback for Control of Reaching and Grasping using Functional Electrical Stimulation

M. Hansen*, K.D. Strange, Y. Chen, and J.A. Hoffer

School of Kinesiology, Simon Fraser University, Burnaby, B.C. V5A 1S6, Canada
*The Center for Sensory-Motor Interaction, Aalborg University, DK-9220, Denmark.

I. Introduction

We investigated, in an animal model, whether sensory feedback from nerve cuff electrodes could provide state control of functional electrical stimulation (FES) to restore reaching and grasping in paralyzed limbs. The approach was to apply threshold-based detection analysis on recorded electroneurographic (ENG) signals to predict recorded electromyographic (EMG) signals from two wrist and digit flexor muscles.

II. Methods

Three cats were trained to perform a reaching and grasping task triggered by an audio signal. The task consisted of reaching for a joystick with the left forepaw, pulling it towards the mouth and holding it (Fig. 1). Once trained, the cats were implanted with tripolar nerve cuff recording electrodes on the radial and median nerves and bipolar EMG muscle patch electrodes in 4-6 muscles of the left forelimb. Implanted muscles were Palmaris Longus (PalL), Flexor Digitorum Profundus (FDP), Extensor Digitorum Lateralis (EDL), Extensor Digitorum Communis (EDC), Flexor Carpi Ulinaris (FCU), and Biceps Brachii (Bic).

Two servomotors independently controlled the joystick fore-aft stiffness and lateral position (Fig. 2). Occasional perturbations in each dimension caused either load changes by increasing the stiffness (X direction in Fig. 1) or slips away from the paw (Y direction in Fig. 1). Cats detected these changes and adjusted their motor output accordingly. The joystick device was fully computer controlled with all parameters being software adjustable.

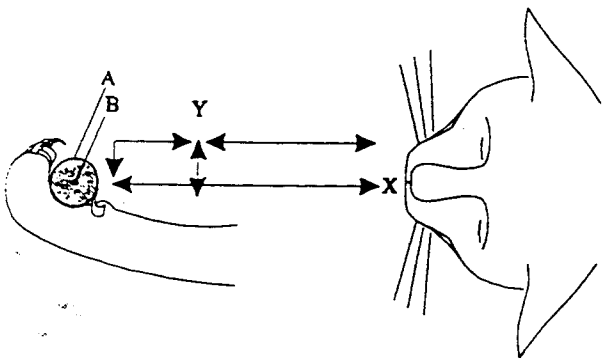

Fig. 1. Top view of movement directions and grasp. Y: movement direction against a virtual spring. Y: slip movement direction. A: lever positioned at the origin. B: food reward delivery tube. As the arrows indicate, the lever returned to its original position if the cat let go after a slip perturbation.

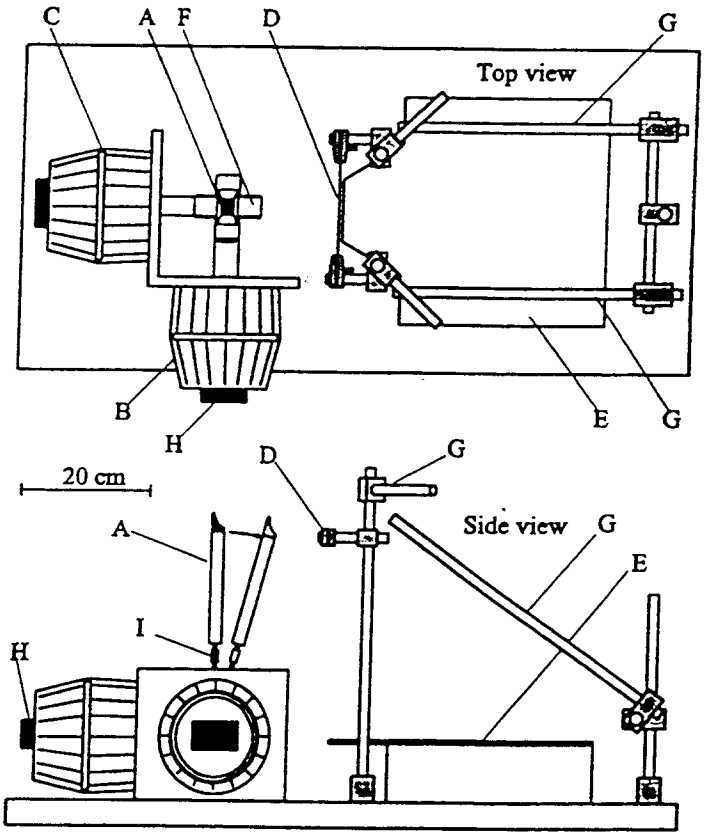

Fig. 2. Joystick hardware. A: lever with food reward tube in it. B: Motor for virtual spring force resistance. C: Motor for inducing slip. D: cat position restraint. E: platform. F: bearing system. G: frame for jacket used to keep the cat on the platform. H: position sensor system. I: strain gauges.

III. Results

EMG recordings showed that for each cat, the individual muscle activation profiles were very consistent and reproducible over trials and over days. Averaged EMG profiles were closely represented by simple piecewise-linear envelope functions that were fitted to averaged data using a Gauss-Newton system identification approach.

ENG signals contained primarily cutaneous information from digits and paw. ENG profiles consistently included recognizable features during phases of reach and grasp and in response to slip perturbations. However, load changes were usually not detectable in the ENG signals with the amplitude and timing of perturbations that were used.

Typical features from normal trials are illustrated in Fig. 3. Px and Py are position of the joystick, Fx and Fy are force applied to the joystick by the cat, Rad and Med are Radial and Median ENG, while PalL, FDP, EDL, EDC, FCU, and Bic are EMG signals. All ENG and EMG signals were bin-integrated into 10ms bins.

An off-line, rule-based threshold detection algorithm was derived that used two sensory ENG channels as input. This provided accurate predictions of actual EMG burst features for two flexor muscles using the derived standard EMG envelope functions (Fig. 4).

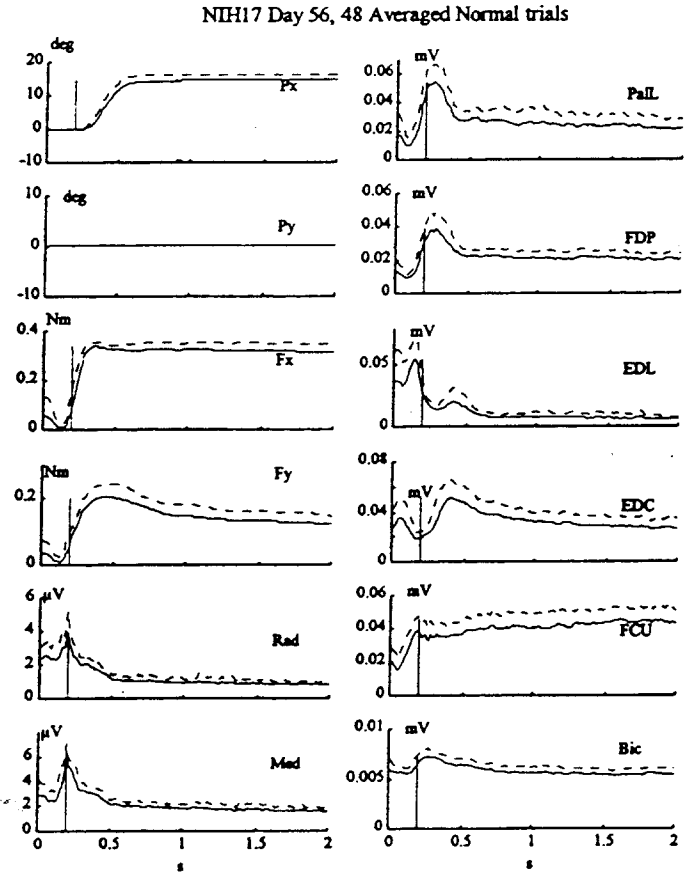

Fig. 3. Averaged signals from 48 normal trials from subject NIH17 on day 56 after implant. Solid curves are averaged signals and dotted curves averaged signals plus standard deviation. The vertical lines in each plot indicate the trigger (contact) point.

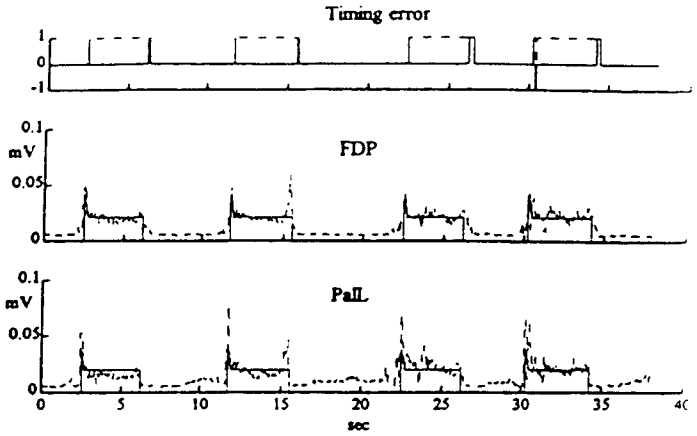

Fig. 4. Top plot is prediction error compared to information from force sensors. The two bottom plots are measured filtered EMG (dotted traces) and predicted scaled EMG activity (solid traces).

The threshold detection algorithm was able to accurately detect paw contact with the joystick in most cases, while detection of the end of the contact phase proved to be more difficult. Attaching the derived envelope functions to the algorithm state variable showed good accuracy when the events were correctly detected from the neural signals.

IV. Discussion and Conclusion

The consistency of the recorded data indicated that cats are easily trained to perform a simple reaching task repeatedly. This consistency proved useful for developing and testing functional electrical stimulation control systems using sensory nerve signals as inputs.

We envision that in human applications of this approach, statistically derived EMG envelope functions may provide templates to synthesize appropriate FES patterns for each paralyzed muscle. In addition, these results also indicate that sensory nerve recordings may be used as feedback sources for real-time control of muscle stimulation onset, timing and amplitude.

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